Joint moments and power in the acceleration phase of bend sprinting

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Abstract

Joint kinetics of the lower limb (hip, knee, ankle, midfoot and metatarsophalangeal joints) were investigated during the acceleration phase of bend sprinting and straight-line sprinting. Within the bend sprinting literature, it is generally accepted that sprint performance on the bend is restricted by moments in the non-sagittal plane preventing the production of force in the sagittal plane. However, there is limited evidence in conditions representative of elite athletics performance that supports this hypothesis. Three-dimensional kinematic and ground reaction force data were collected from seven participants during sprinting on the bend (36.5 m radius) and straight, allowing calculation of joint moment, power and energy. No changes in extensor moment were observed at the hip and knee joints. Large effect sizes (g = 1.07) suggest a trend towards an increase in left step peak ankle plantarflexion moment. This could be due to a greater need for stabilisation of the ankle joint as a consequence of non-sagittal plane adaptations of the lower limb. In addition, the observed increase in peak MTP joint plantar-flexor moment might have implications for injury risk of the fifth metatarsal. Energy generation, indicated by positive power, in the sagittal plane at the MTP and ankle joints was moderately lower on the bend than straight, whilst increases in non-sagittal plane energy absorption were observed at the ankle joint. Therefore, energy absorption at the foot and ankle may be a key consideration in improving bend sprinting performance.

1. Introduction

Research demonstrates a lower velocity during sprinting on the bend than straight (Churchill et al., 2015; Churchill et al., 2016; Judson et al., 2019). At small radii (1-6 m), the need to sustain muscle forces in the frontal and transverse planes is thought to prevent sagittal plane moment generation and inhibit production of ground reaction force (Chang and Kram, 2007). This hypothesis has not yet been confirmed under conditions representative of competitive athletics, yet is generally accepted within the literature at larger radii (Alt et al., 2015; Churchill et al., 2015).

Luo and Stefanyshyn (2012a) utilised wedged footwear, placing the left foot in a more neutral position during sprinting (2.5 m radius), resulting in a lower eversion angle and greater plantar-flexion moment than the control condition. Moreover, greater left limb non-sagittal plane moments were observed during bend sprinting with a weighted vest condition (Luo and Stefanyshyn, 2012b). Therefore, suggesting the left limb is able to generate more force than observed during the control condition, but for some reason is prevented from doing so. However, the radius evaluated (Luo and Stefanyshyn, 2012a) is smaller than those during bend sprinting. Although Viellehner et al. (2016) reported greater left ankle plantar-flexion moment with a 36.5 m radius compared to the straight, the submaximal effort is also not representative of competitive performance. Therefore, a contradiction exists between empirical evidence at smaller radii and submaximal velocity (Luo and Stefanyshyn, 2012a, b; Viellehner et al., 2016), and hypotheses within the literature (Alt et al., 2015; Chang and Kram, 2007; Churchill et al., 2015). Thus, analysis of 3D joint moments at maximal effort and at radii representative of competitive athletics is required.

A proximal-distal sequencing of peak joint extension powers exists during the acceleration phase of straight-line sprinting (Johnson and Buckley, 2001). High peak hip adduction angles during bend sprinting (Alt et al., 2015; Churchill et al., 2015) could impact the ability to produce forces in the sagittal plane and disrupt this proximal-distal sequencing, resulting in a decrease in sprint performance. Furthermore, the complex arrangement of the lower limb and associated non-sagittal plane adaptations could be risk factors for injury during bend sprinting. Sprinters recorded the most muscular injuries in the international athletics championships (2007 - 2015; Edouard, Branco and Alonso, 2016). However, sprint events were recorded as a single category so bend specific injuries cannot be identified. Moreover, iliotibial band syndrome and medial stress syndrome were amongst frequent injuries in collegiate athletes undertaking training on an indoor track (Beukeboom, Birmingham, Forwell and Ohrling, 2000). Analysis of joint kinetics would establish the net demand of the joint and surrounding musculature, thus being influential in developing more specific strength and conditioning and injury prevention programmes. Strength training for sprint performance has focussed upon exercises in the sagittal plane which require triple extension (Young, 2006). Resisted sled training is completed in a straight-line, and plyometric exercises which do adopt a single leg approach make no suggestion of leg-specific exercises (Young, 2006). This is similar to a further review (Wild et al., 2011), suggesting elements of sprint training that comprise a change of direction or are limb specific do not currently occur. Competitive bend sprinting takes place in an anti-clockwise direction, meaning the left limb is always the inside limb, and continued sprinting in the same direction is likely to result in limb-specific differences that warrant individual training interventions. Therefore, this study aimed to investigate lower limb joint kinetics during sprinting on the bend and straight. Conforming to the empirical evidence presented, it was hypothesised that increased frontal and transverse plane moments would be greater on the bend compared with the straight.

2. Methods

2.1. Participants

Following institutional ethical approval, seven male experienced bend sprinters (mean age 22 ± 4 years; body mass 68.32 ± 6.98 kg; stature 1.79 ± 0.06 m; 200 m personal best time: 21.8 - 23.43 s) participated in the study. Participants had no history of injury within the six months prior to data collection and provided written informed consent.

2.2. Experimental set-up

Kinematic data were collected with a 15-camera optoelectronic system (13 x Raptor and 2 x Eagle, Motion Analysis Corporation, CA, USA, 200 Hz). The direction of progression was most closely aligned with the positive *x*-axis, the *y*-axis was vertical and the *z*-axis mediolateral. The calibration volume (7 m long, 3 m wide and 1.5 m high) was located tangentially to the apex of the curve to record data through the 10 – 17 m section of the 30 m sprints (Figure 1). Torso, pelvis, thighs, shanks and feet segments (toebox, forefoot, rearfoot) were modelled using a lower limb and trunk marker set (Judson et al., 2017). Participants were topless and wore form fitting shorts, allowing marker placement directly onto the skin. Foot markers were placed onto the participants' preferred running spike and thought to represent movement of the underlying bones. Force data (1000 Hz) were collected with a Kistler force plate (9287BA, 900 x 600 mm) embedded into the track surface at 11 - 13 m from the start line.

2.3 Protocol

For bend trials, a bend replicating lane 1 (radius 36.5 m) of a standard 400 m running track (IAAF, 2008) was reconstructed on a flat section of indoor track. Simulating competitive bend sprinting, athletes completed bend trials in the anti-clockwise direction (left turn). Participants completed their typical competition warm-up before performing a maximum of six (three left steps and three right steps) 30 m sprints at maximal effort in each condition (bend and straight) in a randomised order. Starting blocks and an '*on your marks, set, go*' signal were used. One researcher modified the location of the start blocks (to a maximum of 1 m forwards or backwards) to aid the athlete in contacting the force plate. Approximately eight minutes were allowed between trials to avoid the onset of fatigue.

2.4 Data processing

Raw 3D marker coordinate data were analysed using Cortex software (version 5.3, Motion Analysis Corporation, CA, USA). Visual 3D (version 6, C-Motion, MD, USA) was used to define and construct segments, local coordinate systems and joint centres in line with ISB guidelines (Wu et al., 2002; Wu et al., 2005). However, the multi-segment foot was defined in accordance with Cappozzo et al. (1995).

Body segment parameters were estimated from de Leva (1996), estimates for a single segment foot were applied to each segment of the multi-segment foot based upon individual segment length (Deschamps et al., 2017; Dixon et al., 2012). Multi-segment foot values were adjusted by 150 to 189 g (according to manufacturer specification) representing the mass of individual participants' spiked shoes (Hunter et al., 2004a) and distributed based on the relative segment length (Bezodis et al., 2012; Bruening et al., 2012). All data were filtered with a low-pass, fourth order recursive Butterworth filter (18 Hz).

One successful trial per condition and per participant was analysed, as with previous sprint research (Johnson and Buckley, 2001). In the current study, analysis of multiple trials was not possible since participants did not always make contact with the force plate more than once. A successful trial was defined as the participant making contact with the centre of the force plate without the presence of targeting. Where multiple successful trials were available, the first successful trial was used for analysis. *Touchdown* and *take-off* were identified from vertical ground reaction force using a threshold mean plus two standard deviations of data where there was zero load on the force plate (Bezodis et al., 2007) All variables were calculated separately for the left and right step. All participants exhibited a foot-strike where the toebox segment was the first point of contact.

Joint moments were calculated in Visual 3D (version 6, C-Motion, Rockville, MD, USA) and expressed in the joint coordinate system, discussed in Schache and Baker (2007). Data were cropped to the propulsive phase of stance and the entire ground reaction force allocated to the toebox segment. Joint moment data were normalised to body weight and height to maintain consistency with sprint literature (Charalambous et al., 2012).

Joint powers in each direction were calculated in Matlab (v2017a, Mathworks, Natick, USA) as the dot product of non-normalised joint moment and joint angular velocity and normalised using the following equation (Hof, 1996), adapted by Bezodis et al. (2010):

$$\frac{\vec{P}}{m \cdot g^{3/2} \cdot h^{1/2}}$$

where $\vec{P}$ is power, m and h are mass and height of the sprinter and g is gravitational acceleration.

2.5 Statistical analysis

Two-way repeated measures analysis of variance (ANOVAs) were performed where condition (bend vs. straight) x limb (left vs. right) were analysed. Effect size (Hedges' g) was used to indicate the magnitude of the effect, interpreted using $g < 0.20$ represents a trivial difference, $0.20 \geq 0.50$ a small difference, $0.50 \geq 0.80$ a moderate difference and ≥ 0.80 a large difference between means (Cohen, 1988).

3. Results

Mean sprint velocities on the straight were 7.96 ± 0.23 m/s (left) and 8.00 ± 0.20 m/s (right), and 7.81 ± 0.30 m/s (left), 7.89 ± 0.34 m/s (right) on the bend. Joint moment and power across the propulsive phase of stance are shown in Figures 1-4. For peak MTP joint plantar-flexor moment, the condition x limb interaction was non-significant ($F_{(1, 6)} = 0.06$, $p = 0.81$). There was a main effect for limb ($F_{(1, 6)} = 8.46$, $p = 0.03$, $g = 1.90$) due to a larger plantar-flexor moment in the right MTP joint than left. Moderate and large effect sizes (g =

0.59, 1.45) suggest a trend towards a greater plantar-flexor moment on the bend than straight in the right and left MTP joints, respectively. A large effect size (*g* = 1.07) suggests a trend towards increased peak left step ankle plantar-flexion moment on the bend compared with the straight, although the interaction was non-significant ($F_{(1, 6)} = 2.33$, $p = 0.18$).

For peak ankle eversion moment, the condition x limb interaction was non-significant ($F_{(1, 6)} = 0.70$, $p = 0.43$). A main effect for limb was observed, $F_{(1, 6)} = 26.00$, $p < 0.01$, due to an increase in left step peak ankle eversion moment on the bend compared with the right step on the bend ($g = 1.23$). A large effect size suggests a greater left step peak ankle eversion moment on the bend than straight ($g = 0.82$). The condition x limb interaction for midfoot eversion was non-significant ($F_{(1, 6)} = 3.20$ $p = 0.12$). However, moderate and large increases in left step peak midfoot eversion moment were observed during bend sprinting compared with the straight ($g = 0.65$) and also when compared with the right step on the bend ($g = 1.31$).

There was no condition x limb interaction ($F_{(1, 6)} = 0.57$, $p = 0.48$) for peak hip flexor moment. Although the main effect for condition was non-significant ($F_{(1, 6)} = 4.47$, $p = 0.08$), a large and moderate ($g = 1.10, 0.70$) decrease in peak hip flexor moment was observed for the left and right steps on the bend compared with the straight, respectively. For peak knee flexor moment a large decrease was observed during the left step of bend sprinting. However, the condition x limb interaction was non-significant ($F_{(1, 6)} = 4.66$, $p = 0.07$). For hip adduction, the condition x limb interaction was non-significant ($F_{(1, 6)} = 2.84$, $p = 0.14$). However, there was an increase ($g = 0.85$) in peak left step hip adductor moment during sprinting on the bend relative to the straight.

3.1 Joint power

At the hip joint, no condition x limb interactions for peak power in the sagittal plane (positive: $F_{(1, 6)} = 0.06$, $p = 0.82$, negative: $F_{(1, 6)} = 0.23$, $p = 0.65$) were observed. Although there was no condition x limb interaction ($F_{(1, 6)}$ =1.59, p = 0.25), there was a trend towards a greater left step peak positive hip power in the frontal plane during (g = 0.82) on the bend than straight (main effect, limb: $F_{(1, 6)}$ = 6.29, p = 0.05). A moderate (g = 0.63) effect size suggests a trend towards greater left step peak negative hip power in the transverse plane during bend sprinting relative to the straight, however no condition x limb interaction was reported ($F_{(1, 6)}$ = 2.88, p = 0.14).

At the ankle, there was no condition x limb interaction for peak negative sagittal plane joint power ($F_{(1, 6)}$ = 0.05, p = 0.841). There was a large, non-significant, increase in left step sagittal plane ankle energy absorption on the bend compared with the straight (main effect limb: $F_{(1, 6)}$ =3.287, p = 0.13, g = 0.80). The condition x limb interaction was non-significant for peak positive transverse plane ankle power ($F_{(1, 6)}$ = 1.13, p = 0.34). However, there was a moderate increase in peak positive left step ankle power in the transverse plane during bend sprinting relative to the straight (main effect limb: $F_{(1, 6)}$ = 4.800, p = 0.08, g = 0.76).

At the MTP joint, no condition x limb interaction was found ($F_{(1, 6)}$ = 0.261, p = 0.627). However, there were large and moderate (g = 0.95, 0.52) increases in peak negative joint power for the left and right step, respectively (main effect condition: $F_{(1, 6)}$ = 6.27, p = 0.05). There was an increase in MTP joint energy absorption (main effect condition: $F_{(1, 6)}$ = 7.14, p = 0.04) on the bend compared with the straight (left, g = 1.68, right, g = 0.30). For midfoot peak positive power, there was a main effect for condition, $F_{(1, 6)}$ = 48.04, p <0.01, with an increase in peak positive midfoot power in both the left (g = 1.20) and right (g = 0.74) step on the bend compared with the straight. For peak negative midfoot power, there was a main effect for limb ($F_{(1, 6)}$ = 19.683, p <0.01, with an increase in peak negative midfoot power in the left step compared with the straight (g = 1.45).

4. Discussion

This study evaluated joint kinetics during the acceleration phase of sprinting on the bend and straight. During bend sprinting, there was a large, but non-significant, decrease in peak flexor moment of the left hip and knee compared with straight-line sprinting. There was a moderate decrease in peak flexor moment of the right hip, although no change was observed at the right knee. Changes in non-sagittal plane moments were also observed, with a trend towards an increase in peak left hip adduction moment.

During both bend and straight conditions, sagittal plane hip moment was extensor for the majority of the propulsive phase of stance, becoming flexor towards the latter stages of stance, similar to early (Bezodis et al., 2014), to mid-acceleration (Johnson and Buckley, 2001), and maximal speed (Bezodis et al., 2008) straight-line sprinting. A large effect size suggests a lower left hip flexor moment during sprinting on the bend than straight. The hip flexor moment towards the end of stance drives the limb forward during the swing phase (Charalambous et al., 2012). Consequently, during bend sprinting athletes may experience difficulties repositioning the left leg due to the kinematic alterations such as greater hip adduction and external rotation observed in Judson et al. (2019*)*. Therefore, swing phase mechanics might increase understanding bend sprinting performance.

Moderate effect sizes suggest a trend towards a greater left step peak negative frontal and transverse plane hip power on the bend than on the straight. This negative power indicates an eccentric contraction of the muscles surrounding the hip, which could stabilise the pelvis during sprinting on the bend (Segal et al., 2009). The pelvis is mutually influenced by each limb, and the left and right limb have been shown to behave differently during bend sprinting with the left limb characterised by a high peak hip adduction angle (Alt et al., 2015; Churchill et al., 2015; Judson et al., 2019). Therefore, a greater level of pelvic control is likely required to overcome these adaptations.

Ankle joint results suggest the limiting factor to sprint performance on the bend is more complex than proposed by Chang and Kram (2007). A greater left step peak plantar-flexor moment was observed on the bend than the straight, supporting results at submaximal effort bend sprinting (Viellehner et al., 2016). These results dispute the idea that moment production at the ankle joint is constrained by moments in the non-sagittal planes. Hunter et al. (2004b) and Johnson and Buckley (2001) suggested ankle plantar-flexor moment acts against anterior rotation of the shank to prevent the collapse of the shank due to the effect of ground reaction force. The increase in left step peak ankle plantar-flexor moment is possibly required to stabilise the shank as a consequence of the non-sagittal plane adaptations of the lower limb. Furthermore, peak MTP plantar-flexor moment was greater during bend sprinting than the straight. MTP moment was plantar-flexor for the duration of the propulsive phase of stance, supporting research in the acceleration phase of straight-line sprinting (Smith et al., 2014; Stefanyshyn and Nigg, 1997). MTP bending moments, attributed to high moments at the forefoot (or toebox) during push-off, are greatest during acceleration movements and a possible risk factor for fifth-metatarsal stress fractures (Orendurff et al., 2009). Bend sprinting likely also increases the torsional load experienced by the long bones of the foot and the increase in ankle and MTP plantar-flexor moments observed in the present study are a risk factor for injury. Orendurff et al. (2009) recommendations for minimising injury risk include careful consideration of the rate at which sprint training volume increases and the number of accelerations performed within a session, and may have particular importance in preparation for bend sprinting.

The ankle joint generated the most power in the sagittal plane (normalised power: 0.97 - 1.24) compared with the hip (0.52 - 0.56) and knee joints (0.09 - 0.12). These findings agree with previous literature demonstrating the dominant role of the ankle joint in sprinting (Brazil et al., 2017; Debaere et al., 2015; Dorn et al., 2012; Johnson and Buckley, 2001).

Large and moderate increases in left step peak positive power were observed at the ankle in the frontal and transverse planes - this did not affect positive power production in the sagittal plane where no differences were observed. Therefore, strengthening foot and ankle muscles in non-sagittal planes specific to bend sprinting may be beneficial in improving bend sprinting performance.

Compared with the hip and ankle joints, power generation at the knee joint was relatively low, suggesting a supporting role at the knee and agree with Heinrich et al. (2015) who proposed the knee may have a sub-unit function, facilitating the transfer of power from the hip through to the ankle. Thus, similar to straight-line sprinting (Bezodis et al., 2008), the knee functions in a supporting role during bend sprinting.

In both conditions, a proximal to distal sequence of peak extensor power generation was observed. However, during bend sprinting with the left step, the timing of peak extensor power at the knee was later than that of the straight. This sequential transfer of power is thought to allow joints such as the ankle to achieve a higher power output despite having relatively smaller muscles (Jacobs et al., 1996). Kinematic adaptations, such as increased left step hip adduction (Alt et al., 2015; Churchill et al., 2015), might delay the transfer of power from the hip to the knee. However, this increase in peak left hip frontal plane power is likely a necessary consequence of the need to generate centripetal force and stay in the correct lane.

The negative sagittal plane power indicates large amounts of energy were absorbed in the ankle and MTP joints in both the bend and straight conditions. Effect sizes suggest a trend towards a greater negative power, and thus energy absorption, in the left MTP, midfoot and ankle on the bend than the straight. During bend sprinting, the left step also demonstrated greater energy absorption than the straight in the frontal plane at the midfoot and ankle, and the transverse plane at the ankle. Therefore, energy absorption in the foot and ankle may be a key consideration for improving bend performance. Smith et al. (2014) demonstrated increased sprint velocity and energy generated at the MTP joint during push-off in a shod condition compared with sprinting barefoot, suggesting sprint spikes are capable of improving MTP joint function, resulting in faster sprint performance. Strength training targeting muscles such as tibialis anterior and posterior, in addition to intrinsic foot muscles such as abductor hallucis and flexor digitorum brevis, might provide a further opportunity to influence sprint performance, particularly on the bend (Smith et al., 2014).

The small sample size of this study is comparable to previous literature (Alt et al., 2015; Churchill et al., 2015; Churchill et al., 2016). Whilst an increased sample size would be desirable in terms of statistical power, the inclusion criteria (200 m PB: 23.5 s) meant this was not possible. Alt et al. (2015) imply velocity specific modulations are apparent during bend sprinting. Consequently, extending the inclusion criteria to include less-skilled sprinters may introduce variability into the sample and was not considered appropriate. Bezodis et al. (2019) suggested differences in sprint start kinematics were more indicative of differences in performance levels, rather than sex. However, to enable the development of specific training interventions, future work investigating the biomechanical adaptations of bend sprinting specific to the female population is warranted. Moreover, the analysis of multiple steps throughout the acceleration phase would be preferable, future research should consider using alternative equipment to allow analysis of a greater number of steps across the acceleration phase. In addition, increased longitudinal bending stiffness in sprint spikes has a localised effect on the MTP joint (Smith et al., 2016). Therefore, different sprint spikes may introduce between-participant differences. Although preferred sprint spikes are warranted to maintain representativeness, future work might examine sprint spike stiffness specifically for bend sprinting performance.

In conclusion, this study demonstrates substantial changes to the function of the joints of the lower limb and loading of the surrounding musculoskeletal structures during bend sprinting compared with straight-line sprinting. Whilst peak flexor moments at the hip and knee were lower during sprinting on the bend than straight, increased plantar-flexor moment at the ankle and MTP suggest the limiting factor to sprint performance on the bend is a complex interaction. Compared with straight-line sprinting, there was also an increase in non-sagittal plane joint moments during bend sprinting, particularly at the hip and ankle joints. To improve bend sprinting performance, athletes should consider developing the ability to produce plantar-flexion from an internally rotated position specific to bend sprinting.

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Conflict of interest statement

The author's declare no conflict regarding the content of this article.

Figure 1. Plan view of the experimental set-up (not to scale).

Figure 2. Hip joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -) and straight (solid line ——). The shaded area indicates standard deviation; for clarity this is presented for the bend trials only.

Figure 3. Knee joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -) and straight (solid line ——). The shaded area indicates standard deviation; for clarity this is presented for the bend trials only.

Figure 4. Ankle joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -) and straight (solid line ——). The shaded area indicates standard deviation; for clarity this is presented for the bend trials only.

Figure 5. Midfoot and MTP joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -) and straight (solid line ——). The shaded area indicates standard deviation; for clarity this is presented for the bend trials only.